Year: 2018

A strain gauge analysis comparing 4-unit veneered zirconium dioxide implant-borne fixed dental prosthesis on engaging and non-engaging abutments before and after torque application

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Abstract: This study quantified the strain development after inserting implant-borne fixed dental prosthesis (FDP) to various implant-abutment joints. Two bone-level implants (= 4.1 mm, RC, SLA 10 mm, Ti, Straumann) were inserted in polyurethane models (N = 3) in the area of tooth nos 44 and 47. Four-unit veneered zirconium dioxide FDPs (n = 2) were fabricated, one of which was fixed on engaging (E; RC Variobase, = 4.5 mm, H = 3.5 mm) and the other on non-engaging (NE) abutments (RC Variobase, = 4.5 mm, H = 5.5 mm). One strain gauge was bonded to the occlusal surface of pontic no. 46 on the FDP and the other two on the polyurethane model. Before (baseline) and after torque (35 Ncm), strain values were recorded three times. Data were analyzed using Kruskal-Wallis and Mann-Whitney U tests (= 0.05). Mean strain values presented significant increase after torque for both E and NE implant-abutment connection type (baseline: E = 4.33 ± 4.38; NE = 4.85 ± 4.85; torque: E = 196.56 ± 188.02; NE = 275.63 ± 407.7; p < .05). Mean strain values based on implant level presented significant increase after torque for both E and NE implant-abutment connection (baseline: E = 4.94 ± 5.29; NE = 5.78 ± 5.69; torque: E = 253.78 ± 178.14; NE = 347.72 ± 493.06; p < .05). The position of the strain gauge on implants (p = .895), FDP (p = .275), and abutment connection type (p = .873) did not significantly affect the strain values. Strain levels for zirconium dioxide implant-borne FDPs were not affected by the implant-abutment connection type.

DOI: https://doi.org/10.1002/cre2.97

Originally published at:
Epprecht, Alyssa; Zeltner, Marco; Benic, Goran; Özcan, Mutlu (2018). A strain gauge analysis comparing 4-unit veneered zirconium dioxide implant-borne fixed dental prosthesis on engaging and non-engaging abutments before and after torque application. Clinical and Experimental Dental Research, 4(1):13-18.
DOI: https://doi.org/10.1002/cre2.97

A Strain Gauge Analysis Comparing 4-Unit Veneered Zirconium Dioxide Implant-Borne Fixed Dental Prosthesis on Engaging and Non-Engaging Abutments Before and After Torque Application

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Abstract

Objectives: This study quantified the strain development after inserting implant-borne fixed dental prosthesis (FDP) to various implant-abutment joints.

Material and methods: Two bone-level implants (BL, ∅:4.1 mm, RC, SLA 10 mm, Ti, Straumann), were inserted in polyurethane models (N=3) in the area of tooth no. 44 and 47. Four-unit veneered zirconium dioxide FDPs (n=2) were fabricated one of which was fixed on engaging (RC Variobase, ∅:4.5 mm, H: 3.5 mm) (E), and the other on non-engaging abutments (RC Variobase, ∅:4.5 mm, H:5.5 mm) (NE). One strain gauge was bonded to the occlusal surface of pontic no. 46 on the FDP, and the other two on the polyurethane model. Before (Baseline) and after torque (35 Ncm), strain values were recorded three times. Data were analyzed using Kruskal Wallis and Mann Witney-U tests (alpha=0.05).

Results: Mean strain values presented significant increase after torque for both E and NE implant abutment connection type (Baseline: E:4.33±4.38; NE:4.85±4.85; Torque: E:196.56±188.02; NE:275.63±407.7) ($p<0.05$). Mean strain values based on implant level presented significant increase after torque for both E and NE implant abutment connection (Baseline: E: 4.94±5.29; NE:5.78±5.69; Torque: E:253.78±178.14; NE:347.72±493.06) ($p<0.05$). The position of the strain gauge on implants ($p=0.895$), FDP ($p=0.275$) and abutment connection type ($p=0.873$) did not significantly affect the strain values.

Conclusions: Strain levels for zirconium dioxide implant-borne FDPs were not affected by the implant-abutment connection type.

Key words: Engaging and non-engaging abutments, FDPs, Oral implants, strain gauge, zirconia

Introduction

The use of dental implants is a well-accepted and predictable treatment modality for the rehabilitation of partially or completely edentulous patients (Hedge, et al. 2009, Asvanund 2014). Although the success rate with implants are high, biological and technical complications around the implants or implant-borne fixed dental prosthesis (FDP) are reported to increase in long-span FDPs (Pjetursson, et al. 2007). The type of implant abutment connection, configurations of implant components[1] or design and biomechanical properties of the FDP material play a significant role on stress distribution around the implants or on the FDP.

Initially, upon tightening, the abutment screw exerts a compressive force to maintain the contact between the abutment and the implant surface (Nishioka, et al. 2010). At this moment, the torque applied to the prosthesis-abutment induces stresses that are transmitted to the supporting bone and suprastructure, which can eventually yield to bone resorption (Nishioka, et al. 2010, Asvanund 2014) or chipping in the veneering ceramic. In fact, mechanical stress may have both positive and negative consequences on the bone tissue (Isidor 2006, Abreu, et al. 2012). While, the response to an increased mechanical stress below a certain threshold may increase the bone density or apposition of bone (Watanabe, et al. 2000, Isidor 2006, De Vasconcellos, et al. 2012), micro-damage as a result of mechanical stress beyond the fatigue threshold results in bone resorption (Watanabe, et al. 2000, Isidor 2006, De Vasconcellos, et al. 2012).

Typically engaging abutments are indicated for crowns but can be used for FDPs as the subtle screw holes are more aesthetic than the larger ones as in the case of non-engaging abutments. In addition, the height (5.5 mm) of the engaging abutment is higher which enables better stability for the framework of the FDP compared to non-engaging (3.5 mm). In practice, the grooves of the engaging abutments are partially eliminated by grinding manually that could result in more strain development in the bone tissue around the oral implants. However, to date, there is no proof whether engaging abutments cause more strain development compared to non-engaging abutments in the FDPs, despite the standardized grinding procedures. Moreover, torque forces during tightening of the prosthetic screws also produces compressive forces on the suprastructure (Asvanund 2014). Depending on the material type, even through the rigid suprastructure appears to fit well on each abutment, residual stresses after the torque may yield to mechanical failures in the FDP (Asvanund 2014).

Complex strain fields around fixtures, implant components or suprastructures could be typically measured using strain gauge analysis (Cehreli & Iplikcioglu 2002, Heckmann, et al. 2004, Karl, et al. 2005, Isidor 2006, Karl, et al. 2008, Hedge, et al. 2009, Abduo, et al. 2011, Karl, et al. 2011, Karl & Taylor 2011, Karl, et al. 2012, Karl & Holst 2012, Nishioka, et al. 2011, Abreu, et al. 2012, De Vasconcellos, et al. 2012, De Vasconcellos, et al. 2013, Asvanund 2014, Cho, et al. 2014, Castro, et al. 2015, Nishioka, et al. 2015). With this method, an electrical resistance in the strain gauge enables the measurement of deformation with high sensitivity (μm/m) (Asvanund 2014). Strain is defined as the ratio between the length of an object under stress and its original dimension; it is a dimensionless entity.[4] In that respect, a strain gauge is considered an indirect measurement method that analyzes mechanical deformation under physical stress, based on electrical measurements registered with a device called a “transducer” (Nishioka, et al. 2010). Since deformations are normally imperceptible to the naked eye, strain gauge is a useful tool as it quantifies a superficial deformation with an electric sensor (Nishioka, et al. 2010). The working principle in this method is based on the variation of the electrical resistance transformed into the deformation levels (Nishioka, et al. 2010). To the best knowledge of the authors, there is no study to date specifically comparing the strain development in engaging versus non-engaging abutment types in relation to the FDP type and the torque amount.

The objectives of this study therefore, were to quantify the strain development after inserting implant-borne FDPs to engaging versus non-engaging abutment types at implant and FDP level before and after torque application. The null hypothesis tested was that the abutment type would not change the strain level at neither implant or FDP level before and after torque application.

Materials and methods

Model preparation

Experimental model was fabricated from a phantom model (Nissin Dental Products Ltd, Kyoto, Japan) where the teeth between 43 and 48 were missing, representing a clinical situation requiring implants. In order to simulate the alveolar bone tissue, models (N=3) were poured in polyurethane (Polyurock, Cendres+Métaux SA, Bienne, Switzerland) having similar mechanical properties to the bone.

Two bone-level implants (BL, ∅: 4.1mm, RC, SLA 10 mm, Ti, Straumann AG, Basel, Switzerland), were inserted in the polyurethane models in the area of tooth no. 44 and 47 with a 5° angle between the implants. Using impression copings, impressions were made with polyether material (Permadyne, 3M ESPE, Minn, USA) and the polyurethane models were copied into plaster models. Then analog implants (BL, RC, Implantat Analog, L 12 mm, Ti, Straumann AG) were inserted in the plaster model that allowed for the fabrication of the FDP.

Fabrication of the veneered zirconium dioxide FDPs

For each model, six identical four-unit zirconium dioxide (Lava, 3M ESPE, Minn, USA) FDP frameworks were made and were subsequently veneered (Creation, Creation Willi Geller International GmbH, Meiningen, Austria) (N=6, n=2 per model) according to the manufacturer`s firing instructions.

On each model, one FDP was fixed on engaging (RC Variobase, ∅: 4.5 mm, H: 3.5 mm, Straumann AG) (E), and the other on non-engaging abutments (RC Variobase, ∅: 4.5 mm, H: 5.5 mm, Straumann AG) (NE). One strain gauge was bonded to the occlusal surface of the FDP on pontic no. 46, and the other two on the polyurethane model, one being distal and the other mesial to the two implants (Figs. 1a-b).

While the intaglio surfaces of the FDPs and the metal abutments were air-borne particle abraded with 50 μm silica particles coated with Al_2O_3 (Rocatec Plus, 3M ESPE). Both the FDP and the abutments were ultrasonically cleaned (Bransonic Ultrasonic Cleaner 3510, Branson, Danbury, USA) in ethanol for 5 min and dried oil free air. Then one coat of silane (ESPE-Sil, 3M ESPE) was applied on the FDP and the abutment, waited for its reaction for 5 min. Finally, FDPs were cemented on the abutments using chemically polymerized resin cement (Panavia 21, Kuraray GmBH, Tokyo, Japan). The margins of the FDP were coated with the oxygen blocking gel (Oxyguard, Kuraray GmbH) for 5 min. Then it was washed and dried with oil-free air.

Strain gauge analysis

In order to position the strain gauges (SG) precisely on the polyurethane models, a line connecting the two implants was drawn with a ruler and a 0.7 mm pencil lead. One SG was placed distally adjacent to the implant no. 44 and mesially adjacent to implant no. 47. The third SG was placed on the occlusal surface of pontic no. 46 on the FDP. For exact positioning of SG on the FDP, occlusal surface was made plane. A mesio-distal line was drawn occlusal, leading exactly through the middle of the pontic 46. The SG was placed on this line, bordering right on the edge of the pontic (Fig. 2a).

The sites were initially cleaned with acetone to ensure good bonding of the SGs. A thin layer of methyl-2-cyanacrylate resin (M-Bond 200; Vishay Measurements Group, Raleigh, NC, USA) was used to fix each SG, which was positioned and held in place under slight pressure for three minutes. Soldering terminals were bonded next to the SGs in the same manner. Each SG was wired separately and the three SGs were connected to a multichannel bridge amplifier to form one leg of the bridge. A computer was interfaced with the bridge amplifier to record the output signal of the polyurethane and suprastructure surface. Data-acquisition system software (SignalExpress, National Instruments, National Instruments Corporation, Austin, TX, USA) was used to record the data (Fig. 2b).

Each SG was set to zero and calibrated prior to insertion of the FDPs to the implant. Baseline values were noted at 0 s, 20 s and 40 s after calibration. The occlusal screws were tightened onto the abutments until the screw came to a halt. A torque of 35 Ncm was then applied using the manufacturer's manual torque-controlling device (Straumann AG). Strain was measured again at 0 s, 20 s and 40 s after torque application.

Statistical analysis

Data were analyzed using a statistical software package (SPSS Software V.20, Chicago, IL, USA). Kolmogorov-Smirnov and Shapiro-Wilk tests were used to test normal distribution of the data. As normal distribution was not observed, the data were analysed using Mann Whitney U test and Kruskal Wallis non-parametric tests where strain values were the dependent variables and implant position (2 levels: 44 vs 47), abutment type (2 levels: engaging versus non-engaging) and measurement time (2 levels: baseline versus after torque) were the independent variables. Bonferroni correction was made at $p<0.0083$. P values less than 0.05 were considered significant in all tests.

Results

Overall, compared to baseline, regardless of the implant and FDP level, mean strain values presented significant increase after torque for both E and NE implant abutment connection type (Baseline: E: 4.33±4.38 ; NE: 4.85±4.85 ; Torque: E: 196.56±188.02; NE: 275.63±407.7) ($p<0.05$) (Table 1).

Considering only the strain values on the FDP level, also a significant increase was observed for both E and NE (Baseline: E: 4E: 3.11±1.9 ; NE: 3.0±2.31 ; Torque: E: 82.11±64.22; NE: 131.44±101.74) ($p<0.05$) (Table 2).

Mean strain values based on implant level, presented significant increase after torque for both E and NE implant abutment connection type (Baseline: E: 4.94±5.29 ; NE: 5.78±5.69 ; Torque: E: 253.78±178.14; NE: 347.72±493.06) ($p<0.05$) (Table 3).

The position of the strain gauge being on implant (no.44 and no. 47) (P=0.895), FDP no.46) (P=0.275) and abutment connection type (P=0.873) did not significantly affect the strain values. Strain levels for 4-

unit veneered zirconium dioxide implant-borne FDPs were not affected by the implant-abutment connection type on the model tested.

Discussion

This study was undertaken in order to evaluate the strain development after inserting implant-borne zirconia FDPs to engaging versus non-engaging abutment types at implant and FDP level before and after torque application. Based on the results obtained, since abutment types and position of the implants did not significantly affect the strain development, the first part of the null hypothesis could be accepted. After torque application however, strain values increased significantly in all conditions. Thus, the null hypothesis on torque effect could be rejected.

A number of factors might influence strain development at the implant and FDP level such as the effect of axial and non-axial loading (Nishioka, et al. 2010, Nishioka, et al. 2011, Cho, et al. 2014), straight and offset implant placement (Nishioka, et al. 2010, Cho, et al. 2014), impression technique, fabrication method of the FDPs, retention type and ceramic veneering (Cehreli & Iplikcioglu 2002, Karl & Taylor 2011), the type of implant-abutment joint (Hedge, et al. 2009, Karl, et al. 2008, Cho, et al. 2014) and the type of prosthetic coping (Abreu, et al. 2012). Since no data is available in the current literature focusing on the difference in strain development between engaging and non-engaging abutments, a direct comparison with the other studies would not be possible. Yet, in general it is commonly accepted and widely reported that after torque application, strain levels increase at both implant[2] and FDP level (Nishioka, et al. 2010)

Principally, the cervical region of the implant is the site where the highest stresses occur (Watanabe, et al. 2000, Karl & Holst 2012). This phenomenon is due to the fact that when two materials are in contact with each other and one of them is loaded, the stresses will be higher at the first point of contact in any material (Nishioka, et al. 2015). Therefore, the cervical region of the implant is the site where the greatest microdeformations occur, regardless of the type of bone, the design of the implant, the configuration of the prosthesis, and the load (Watanabe, et al. 2000, Karl & Holst 2012). Hence, in this study, the strain gauges were bonded adjacent to the implant on the polyurethane block through which strain development has been measured by means of SGs. The model could not allow positioning the SGs at the buccal aspect of the implants due to non-flat surfaces. A similar manner of SG positioning was practiced in previous studies, noting that the models were obtained from a real patient case. A standardized model could have allowed such a positioning. Nevertheless, in multiple unit FDPs, deflection occurs mesial and distal of the abutments during loading (Watanabe, et al. 2000, Karl & Holst 2012). Thus, strain values from the regions are of clinical relevance. Moreover, technically, these regions presented completely flat surfaces enabling accurate bonding of the SGs.

In an attempt to simulate the alveolar bone, polyurethane blocks were used in this study similar to numerous previous studies (Watanabe, et al. 2000, Karl, et al. 2008, Abreu, et al. 2012, De Vasconcllos, et al. 2012, Cho, et al. 2014, Nishioka, et al. 2015). Even though, the use of this material is common practice in strain analysis in implantology, polyurethane is assumed to be linearly elastic and isotropic, meaning that the material has the same mechanical properties in all direction. In turn, bone is anisotropic, contains voids and quality of the bone varies as a function of many other factors (Karl, et al. 2008). Nevertheless, the polyurethane model eliminates possible confounding factors related to the biological bone substance.

The results of the current study presented no significant difference in strain development between engaging and non-engaging implant abutment connections. It has to be however noted that after inserting the FDPs in the polyurethane models and applying a torque of 35 Ncm, more strain development was observed in the non-engaging abutments compared to the engaging ones. Yet, the results were not significant. This is most probably due to the limited sample size and therefore this study should be considered as a pilot one. Certainly, impression methods and the duplication procedures of multiple models, adds to the misfit of the FDP (Karl, et al. 2008). Similarly, the transfer of implant position from polyurethane to plaster models could increase errors that eventually affect the accuracy of the measurement between each model. However, this inherent error is valid for both the implant and the FDP and maybe less in a clinical scenario where single duplication is needed.

Nonetheless, the engaging abutments contain insertion grooves in order to avoid undesired rotational movement within the implant whereas the non-engaging one does not. Insertion grooves are beneficial elements in engaging abutments in order to avoid rotation of single crowns while this becomes less of an issue in an FDP. On the other hand, even though the favourable height of engaging abutments for crowns may be also useful for the FDPs compared to non-engaging abutments. Unfortunately, insertion grooves on engaging abutments do not allow easy path of insertion for the FDPs and thus, they were adjusted manually using rotating instruments. One reason for the increased tendency for high strain formation with the non-engaging ones which are typically indicated for crowns as opposed to non-engaging ones, being indicated for FDPs, could be due to the configuration differences, namely non-engaging implant abutment connections present a larger screw hole but less height than that of engaging ones. Hence, it can be anticipated that less height of the non engaging abutment did not support the framework and the veneering ceramic compared to engaging one and consequently after torque, unsupported areas in the FDP caused more stress and thereby more strain development with this abutment both in FDP and implant level.

The results of this study should be verified in a larger sample, noting that they are costly studies. This pilot study allowed us to calculate the power for similar future studies in that 26 specimens are needed with relevant difference of 80mV and standard deviation of 100mV between groups at 80% certainty based on two-sided two-sampled t-test.

Conclusions

Engaging or non-engaging abutments presented similar strain development at both implant and FDP level after torque application compared to baseline measurements. Strain levels at the implant level were higher than on the FDP, yet being not significant. Both engaging and non-engaging abutments could be advised in conjunction with 4-unit veneered zirconia FDPs as they demonstrated similar increase in strain development after torque application.

Disclosure

The authors claim to have no financial interest, either directly or indirectly, in the products or information listed in the paper.

ACKNOWLEDGMENTS and source of funding

The authors would like to acknowledge Mr. A. Trottmann, Mr. T. Di Bello, Mr. D. Reichart, Mr. G. Voce, University of Zurich, Center for Dental and Oral Medicine, Zürich, Switzerland, for their assistance with the fabrication of the FDPs, Mr. S. Erni from the Clinic of Masticatory Disorders, Removable Prosthodotnics, geriatric and Special Care Dentistry for his assistance with the strain gauge analysis, and Dr. M. Roos from the Division of Biostatistics, Institute of Social and Preventive Medicine, University of Zurich, Switzerland for her support with the statistical analysis.

The authors have no support or funding to report.

Conflict of Interests

The authors declare that they have no conflict of interests related to this study.

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Captions to the tables and figures:

Tables

Table 1. Strain values (mV) (Mean±standard deviation), maximum, minimum and confidence intervals (95%) at baseline, after torque and difference between baseline and torque. Different upper-case letters in each column for each condition indicates significant differences ($p<0.05$). E: Engaging; NE: Non-engaging.

Table 2. Strain values (mV) (Mean±standard deviation), maximum, minimum and confidence intervals (95%) at baseline, after torque and difference between baseline and torque considering the FDP 46 occlusal only. Different upper-case letters in each column for each condition indicates significant differences ($p<0.05$). E: Engaging; NE: Non-engaging.

Table 3. Strain values (mV) (Mean±standard deviation), maximum, minimum and confidence intervals (95%) at baseline, after torque and difference between baseline and torque considering the Implant 44 distal; Implant 47 mesial only. Different upper-case letters in each column for each condition indicates significant differences ($p<0.05$). E: Engaging; NE: Non-engaging.

Figures

Figs. 1a-b. Photos of (a) non-engaging (h: 3.5 mm) and (b) engaging implant abutment connection (h: 5.5 mm).

Figs. 2a-b. Position of strain gauges (a) placed distally adjacent to implant no. 44, mesially adjacent to implant no. 47 and on occlusal surface of pontic 46 on the FDP, (b) Soldering terminals placed directly next to the straing gauges that are connected to a multichannel bridge amplifier.

Tables:

Baseline						
Implant Abutment Connection Type	**n**	**Mean (SD)**	**Minimum**	**Maximum**	**Confidence Interval**	
					Lower Bound	Upper Bound
E	9	4.33 ± 4.38^{A}	0.33	14.00	0.96	7.70
NE	9	4.85 ± 4.85^{A}	0.33	16.33	1.13	8.58

After Torque						
Implant Abutment Connection Type	**n**	**Mean (SD)**	**Minimum**	**Maximum**	**Confidence Interval**	
					Lower Bound	Upper Bound
E	9	196.56 ± 168.02^{A}	8.00	448.67	67.40	325.71
NE	9	275.63 ± 407.7^{A}	16.00	1333.67	-37.76	589.02

Difference						
Implant Abutment Connection Type	**n**	**Mean (SD)**	**Minimum**	**Maximum**	**Confidence Interval**	
					Lower Bound	Upper Bound
E	9	192.22 ± 164.84^{A}	7.00	443.33	65.51	318.93
NE	9	276.19 ± 407.61^{A}	15.67	13332.67	-37.13	589.50

Table 1. Strain values (mV) (Mean±standard deviation), maximum, minimum and confidence intervals (95%) at baseline, after torque and difference between baseline and torque. Different upper-case letters in each column for each condition indicates significant differences ($p<0.05$). E: Engaging; NE: Non-engaging.

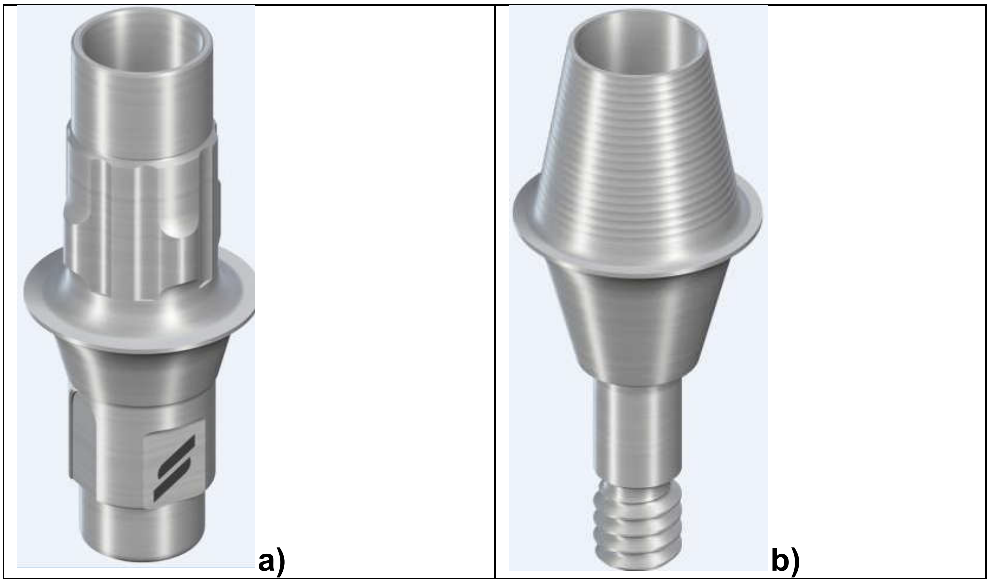
Baseline						
FDP	**n**	**Mean (SD)**	**Minimum**	**Maximum**	**Confidence Interval**	
					Lower Bound	Upper Bound
E	9	3.11±1.90^{A}	1.00	4.67	-1.60	7.82
NE	9	3.00±2.31^{A}	0.33	4.33	-2.74	8.74
After Torque						
FDP	**n**	**Mean (SD)**	**Minimum**	**Maximum**	**Confidence Interval**	
					Lower Bound	Upper Bound
E	9	82.11±64.22^{A}	8.00	121.33	-77.42	241.64
NE	9	131.44±101.74^{A}	16.00	208.00	-121.28	384.17
Difference						
FDP	**n**	**Mean (SD)**	**Minimum**	**Maximum**	**Confidence Interval**	
					Lower Bound	Upper Bound
E	9	79.00±62.38^{A}	7.00	116.67	-75.95	233.95
NE	9	128.44±99.47^{A}	15.67	203.67	-118.65	375.54

Table 2. Strain values (mV) (Mean±standard deviation), maximum, minimum and confidence intervals (95%) at baseline, after torque and difference between baseline and torque considering the FDP 46 occlusal only. Different upper-case letters in each column for each condition indicates significant differences ($p<0.05$). E: Engaging; NE: Non-engaging.

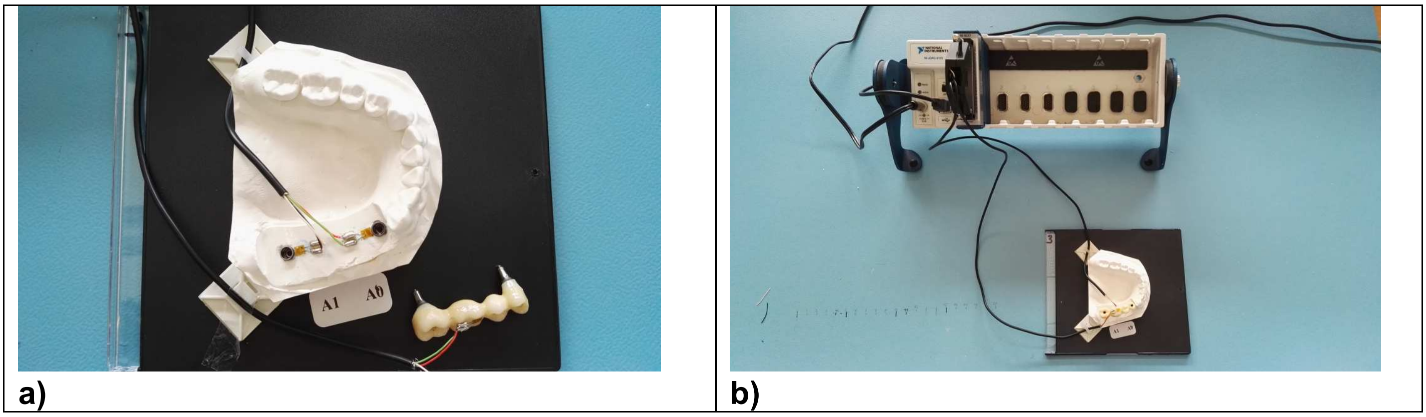
FDP	n	Mean (SD)	Minimum	Maximum	Confidence Interval	
					Lower Bound	Upper Bound
Baseline						
E	9	4.94±5.29^{A}	0.33	14.00	-0.61	10.50
NE	9	5.78±5.69^{A}	1.00	16.33	-0.19	11.75
After Torque						
FDP	**n**	**Mean (SD)**	**Minimum**	**Maximum**	**Confidence Interval**	
					Lower Bound	Upper Bound
E	9	253.78±178.14^{A}	25.00	448.67	66.83	440.72
NE	9	347.72±493.06v	52.33	1333.67	-169.71	865.15
Difference						
FDP	**n**	**Mean (SD)**	**Minimum**	**Maximum**	**Confidence Interval**	
					Lower Bound	Upper Bound
E	9	248.83±174.31^{A}	24.67	443.33	65.91	431.76
NE	9	350.06±492.17^{A}	48.33	1332.67	-166.44	866.55

Table 3. Strain values (mV) (Mean±standard deviation), maximum, minimum and confidence intervals (95%) at baseline, after torque and difference between baseline and torque considering the Implant 44 distal; Implant 47 mesial only. Different upper-case letters in each column for each condition indicates significant differences ($p<0.05$). E: Engaging; NE: Non-engaging.

Figures:



Figs. 1a-b Photos of (a) non-engaging (h: 3.5 mm) and (b) engaging implant abutment connection (h: 5.5 mm).



Figs. 2a-b Position of strain gauges (a) placed distally adjacent to implant no. 44, mesially adjacent to implant no. 47 and on occlusal surface of pontic 46 on the FDP, (b) Soldering terminals placed directly next to the straing gauges that are connected to a multichannel bridge amplifier.